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TITLE: Method of correcting for magnetic field inhomogeneity in magnetic resonance imaging

Abstract Text (1):

A method of correcting for magnetic field inhomogeneity caused by various factors, such as implanted metal and air/tissue interfaces, in magnetic resonance imaging (MRI) is provided. Geometric distortion due to inhomogeneity in a static magnetic field B.sub.0 is corrected for by addition of a compensation gradient. The compensation gradient is applied in the **slice selection** direction Z, has a timing substantially identical to the standard frequency encoding gradient G.sub.x, and has an amplitude identical to the **slice selection** gradient G.sub.z that is applied during the initial RF **excitation**. Inhomogeneity in an RF field B.sub.1 is compensated for by utilizing an RF coil that is large enough in size as compared with a metal implant to make the volumetric percentage of the metal in the coil insignificant. Inhomogeneity in a gradient field G=(G.sub.x, G.sub.y, G.sub.z) is corrected for by a treatment of the most significant error factor G.sub.z that causes **slice** thickness error. Specifically, the method **acquires** two images with complementary **slice** thickness error by using two pulse sequences with flipped **slice selection** gradients G.sub.z ; combination of the two images successfully cancels the effect of the **slice** thickness error. **Local** dephasive MRI signal loss due to magnetic field inhomogeneity is corrected for by **acquisition** of two images with positive and negative offset G.sub.z gradient lobes, respectively. The pair of images are combined to cancel the effect of **local** signal loss error.

Brief Summary Text (6):

The present invention provides a novel method of correcting magnetic field inhomogeneity caused by various factors, such as implanted metal and nonmetallic devices and air/tissue interfaces. The method treats distortions caused by inhomogeneity in all three magnetic fields involved in MRI: a static magnetic field B.sub.0 to polarize the sample, an RF field B.sub.1 for signal **excitation,** and a vector gradient fieldg G=(G.sub.x,G.sub.y,G.sub.z) for spatial encoding. First, the method

corrects for geometric distortion due to inhomogeneity in the static magnetic field B.sub.0 by modifying the magnetic field gradients that are used during the imaging pulse sequence. Specifically, the method adds to a standard pulse sequence a compensation gradient waveform that is applied in the **slice selection** direction (Z), has a timing substantially identical to the standard frequency encoding gradient (G. sub.x), and has an amplitude identical to the **slice selection** gradient (G.sub.z) that is applied during the initial RF **excitation**. Second, the method corrects for inhomogeneity in the RF field B.sub.1 by utilizing an RF coil that is large enough in size, as compared with a metal implant, to make the volumetric percentage of the metal in the coil insignificant. Third, the method corrects for inhomogeneity in the gradient field G=(G.sub.x,G.sub.y,G.sub.z) by treating the most significant error factor, (G.sub.z), by a novel method of **slice** thickness error correction. Specifically, the method **acquires** two images using two pulse sequences that are identical to each other except that their **slice selection** gradients are opposite. Since the **slice** thickness errors in the two images are complementary to each other, the method combines the two images to successfully cancel the effect of the **slice** thickness error. Lastly, the method includes treatment of **local** dephasive MRI signal loss, due to magnetic field inhomogeneity, by **acquiring** two images with positive and negative offset G.sub.z gradient lobes, respectively. The pair of images are combined to cancel the effect of **local** signal loss error.

Detailed Description Text (5):
FIG. 1A illustrates **one** exemplary use of the compensation gradient of the invention in a standard spin-echo sequence. Referring to FIG. 1A, a standard spin-echo **pulse** sequence begins with the transmission of 90. degree. RF **pulse**. As is known in the art, at the termination of the initial 90.degree. **pulse,** the magnetic moments of the individual nuclei are precessing around the Z axis within the X-Y plane. Application of a slice selection gradient G.sub.z **simultaneously with the RF pulse** causes the nuclear spins of a narrow slice in the imaged object along an X-Y plane to be **excited** into resonance. After the application of the 90. degree. RF **pulse,** a 180.degree. RF **pulse** is applied, which has the effect of rephasing the spins to produce a spin echo signal. This spin echo signal is acquired during a readout gradient G.sub.x. Also as is known in the art, a frequency encoding gradient G.sub.x is applied after the 90.degree. RF **pulse,** but before the readout gradient G.sub.x, to **center** the spin echo signal within the readout gradient G.sub.x. The above sequences are repeated with **different phase** encoding gradients G. sub.y, as is known in the art, to acquire multiple data sets from which the imaged object may be reconstructed.

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